AN ANALYTICAL MODEL OF A SPINAL ROOT STIMULATED NEUROMUSCULAR SYSTEM

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Abstract- **Leg Cycle Ergometry using Functional Electrical Stimulation provides many health benefits for paraplegic users. The Lumbo-sacral Anterior Root Stimulator Implant provides sufficient control over the muscles in the legs for leg powered cycling to be achieved, however for further development to take place, more knowledge is needed about the complicated pathways involved in the neuromuscular system from the nerve roots to the muscles in the leg. To achieve this, an analytical model has been developed from experimental measurements which predicts the force output from the leg due to stimulation on one of the twelve available channels dependent on stimulation level, duration of stimulation and leg position (crank angle). The model incorporates both linear and non-linear elements and is a first step towards a full model of the complicated non-linear processes involved in the neuromuscular system. The work here presents both the measurements used to develop the model and the modelled force response**

Keywords - **Functional Electrical, Stimulation, Nerve Root, Cycling, Paraplegic**

I. Introduction

The use of functional electrical stimulation (FES) leg cycling ergometry (LCE) provides many health benefits for paraplegic subjects, including improved cardiopulmonary capacity, muscle bulk and skin tone [1]. However, for paraplegic subjects to wish to use LCE as a regular tool for exercising, FES-LCE systems must be designed that are convenient and enjoyable to use. For these reasons, one of the most important factors in any FES-LCE system is the ability to enable the user to maintain a fluid cycling gait, i.e. the legs do not stop or slow significantly during the revolution. Absence of a fluid cycling gait often results in the user spending a large proportion of time assisting the legs with their hands. This requires extra effort and reduces the exercise time accordingly.

To produce a fluid cycling gait, the leg muscles must be stimulated in a suitable sequence which produces sufficient force or momentum throughout each revolution. Momentum can be maintained in regions where force is not applied by using a flywheel such as in the Ergys series of cycling ergometers. However, if locomotive cycling is to be achieved, a propulsive force must be provided by the legs for the majority of each revolution.

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The Lumbo-sacral Anterior Root Stimulator Implant (LARSI) [2] was designed in an attempt to enable paraplegic subjects to stand with sufficient balance such that they could remove one hand from their standing frame to perform reaching and grasping functions. In order to stimulate the majority of the muscles in the legs while minimizing the extent of the surgical procedure, the implant stimulates 12 motor nerve roots in the lumbar and sacral regions of the spine. These are the anterior roots L2 to S2 bilaterally.

Using the LARSI system we have sufficient control over the muscles in the legs of our first subject to make her produce a fluid cycling gait for locomotive cycling [3]. However, the complicated nerve pathways between the stimulated nerve roots and the muscles in the leg are not fully understood. In order to enhance the cycling capability of our subject, more knowledge of the multidimensional mapping between stimulation and leg response is needed.

In this work we present an analytical model which describes the response map between the stimulation applied to one nerve root in our subject and the resulting force applied to the tricycle pedal, in the direction of motion, by her foot. We also present the experimental measurements upon which this model is based.

II. Methodology

A. Measurement Procedure

Measurements were taken on three occasions over a period of five months. The first session was used to prove the technology while data from the last two sessions is given below. Each test session began with calibration of the systems that measure the crank position and pedal force [4]. Once this was complete, our subject transferred onto the tricycle using a slider board with assistance from one person to support the weight of her legs. Her feet and lower legs were then strapped into orthoses mounted on the pedals. After our subject had indicated that she was ready and comfortable, the stimulation test sequence was started.

Stimulation consisted of a series of short pulse trains. A pause of one second was used between pulse trains to allow the legs to relax to their resting position. The maximum test series length was 60 seconds. This prevented the muscles from fatiguing during a test session. The stimulation pulse rate was 20Hz and pulse trains contained between one and twelve pulses. During this study only one of the twelve possible channels was stimulated. This was the fourth anterior lumbar nerve root on the left (L4L) and

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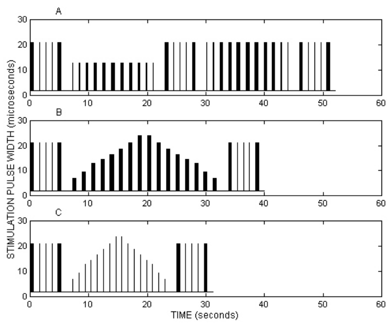

Fig. 1: Test stimulation sequences to examine (A) the effect of pulse train duration between 1 and 12 pulses, represented by the width of the bar. (B) effect of stimulation pulse width on the force response with 12 pulses in each pulse train and (C) the effect of stimulation pulse width on the single pulse responses. The effect of crank angle on the force response was found by repeating the test sequences at a number of crank angles.

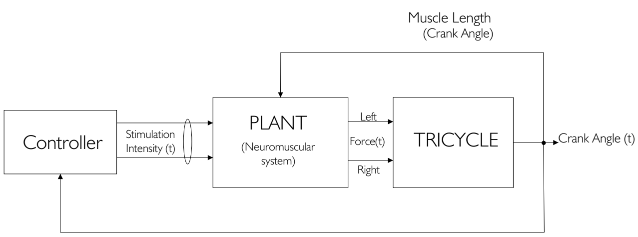

Fig. 2: Proposed system description.

was chosen as it is currently used as our subject's main left knee extensor during cycling. The results presented here were obtained using the stimulation sequences shown in fig 1.

The stimulator uses a constant stimulus current, fixed at an estimated 3.2mA. Modulation is by stimulus pulse width; for these tests this ranged from 7μs to 24μs. This range has been found, through previous testing using the multi-moment chair system [5], to give useful knee extension on L4L.

Measurements were made using a laptop PC with an analogue-to-digital converter (ADC) module attached to the receiver of a force telemetry system whose transmitter was mounted on the pedal cranks of the tricycle. The sample rate of the ADC was set at 50Hz and the position of the pedal was simultaneously acquired from a shaft encoder on the tricycle.

B. Model Development

We start by considering the neuromuscular system between the nerve root L4L and the leg of our subject to be the plant we wish to model, fig.2. We know the input to the plant which is the stimulation signal from the controller, and we measure the useful output from the plant in the form of the force applied to the pedals tangential to the direction of crank rotation.

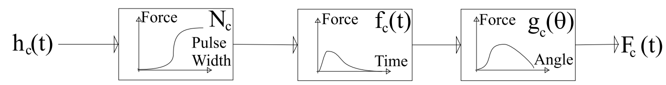

Fig. 3: A block diagram showing the processes involved in the plant we wish to model. The plant involves the neuromuscular system between an electrode attached to our subject's left 4th anterior lumbar nerve root and her left foot.

Next we take into account the known factors that affect the force response of a muscle when stimulated by FES. If we do not continue to stimulate after fatigue has set in then the main factors affecting response are stimulation pulse width and amplitude [6], interpulse period [7], duration of stimulation pulse train, muscle length and electrode placement [8].

In the LARSI system the stimulation amplitude and the interpulse period are held constant. Also, the electrode positions have not changed significantly in our subject over the past six years. We therefore neglect these three factors in our model.

The force output measured by the crank telemetry system is the force applied by the foot to the pedal in a direction that is tangential to the motion of the pedal. This is a transformation of the resultant force produced by all of the muscles that are stimulated in the leg [4]. It can be considered to be the useful (propulsive) force output. When using the force telemetry system it is not possible to determine the effects of each individual muscle length nor is it possible to determine the total force produced by the leg. However, further simplification of the model comes from noting that each foot is only able to travel through a fixed locus because it is attached by an orthosis to the pedal and therefore both the lengths of the muscles and the force transformation vary simultaneously as the crank rotates. We therefore group these factors into a single measurable function based on the crank angle

The peak force produced by the leg is related to the stimulation pulse width by a nonlinear transformation. This transformation is required because of the way in which the nerve fibers are recruited by the stimulation. We assume that this nonlinear transformation is independent of crank angle, θ.

The plant dynamics were determined from the response to a step input. We assumed the dynamics to be of the type normally associated with an over-damped second order differential equation. Once the dynamic system could be described, the impulse response of the plant was derived for use in the model.

C. Model Description

For any given channel, c, the processes involved in the plant are shown in fig.3. The response of the plant, $F_c(t)$, over time, t, is described by equation.1.

$$F_c = g_c\big(\theta(t)\big) f_c(t) * \mathbb{N}\big(h_c(t)\big) \qquad (1)$$

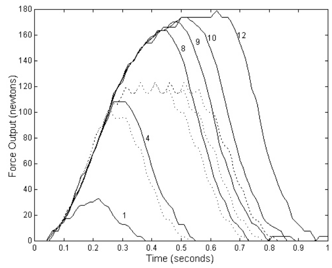

Fig. 4: The effect of the number of pulses in the pulse train on the force output. Note that steady state output starts after 10 stimulation pulses have been fired. The number of pulses for each train is shown on the plot, solid lines are measured data and dotted lines show the effect of linearly summating the single pulse response

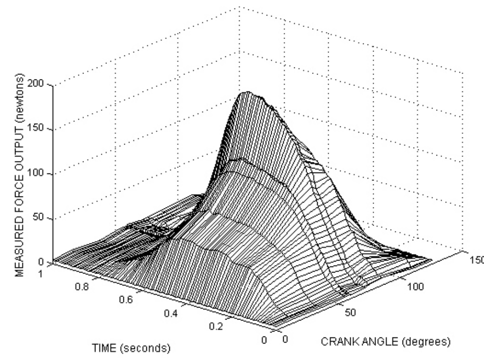

Fig. 5: The effects of the angle of the crank on the waveform of the output force when stimulated with pulse trains of 10 pulses duration

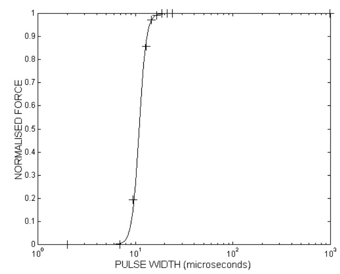

Fig. 6: A plot of the nonlinear mapping, $\mathbb{N}_{L4L}$, between the pulse width of the physical stimulation and the normalized peak force response for stimulation of the 4th anterior lumbar nerve root on the left (L4L)

where $h_c(t)$ is a representation of the stimulation applied to the plant and $\mathbb{N}\big(h_c(t)\big)$ describes the non-linear transformation of $h_c(t)$ due to the physical process by which the nerve fibers are recruited by the stimulation. $f_c(t)$ is the impulse response of the plant derived from the plant dynamics and $g_c\big(\theta(t)\big)$ is the effect of the angle of the crank, $\theta(t)$, on the plant response.

The physical stimulation of the neuromuscular system consists of discrete rectangular pulses with varying width, constant amplitude and constant repetition period of 50ms. However, in our model the input sequence, $h_c(t)$, will be regarded as a piecewise continuous variable with each physical stimulation pulse being represented by a rectangular pulse of 50ms duration with amplitude that is dependent on the pulse width of the physical stimulation.

In this model we assume that the dynamic system is linear and that the time constants of f_c are independent of θ

III. Results

A. Measured dynamic force responses

The effect of the number of pulses in the pulse train can be seen in fig.4. This data was measured while using test sequence A in (fig.1). The measurements are very repeatable. However if we take the response to one pulse in the pulse train and use temporal summation to approximate a series of pulses in the train, the resulting curve does not closely resemble the measured response for longer pulse trains. This indicates that the dynamics are not entirely linear. It is also noted that a large portion of the rise of the force response has a constant gradient. This shape suggests that the response is limited by a maximal rate of change analogous to the slew rate of transistors.

Fig.5 shows the effects of crank angle on the force waveforms that are produced with stimulation pulse trains of 10 pulses duration. The maximum force occurs at around 75 degrees after the angle at which the left hip is most flexed. The measurements show that the dynamics are somewhat dependent on crank angle. Again the results show a maximal rate of increase of force which is constant across the angular range. However rather than the relaxation rate being constant, it appears that the relaxation time is constant across the angular range, that is, the amount of time for the leg to relax to its resting state after stimulation has stopped appears to be independent of angle.

The effect of changing the stimulation level is shown in fig.6. A pulse width of 7μs does not activate the muscles. Initially, increases in stimulation intensity result in a rapid rise in maximal force output. However, after a pulse width of 19μs is reached, further increases in stimulation intensity give little or no further increase in force response.

B. Curve fitting

Using our measurements it has been possible to determine a model for the stimulation channel $c = L4L$. A 6th order polynomial fit was found for the effect of crank angle on the response, g_{L4L}, and a sigmoid function was used to model the nonlinearity, $\mathbb{N}_{L4L}(I_{L4L})$ as shown in fig.6. The equation for this nonlinearity is

$$\mathbb{N}\big(h_{L4L}(t)\big) = \frac{1}{1 + e^{\left(\frac{-0.75 \times (log_2(h_{L4L}/2)}{0.0894} + 20.5\right)}} \qquad (2)$$

The normalised step response of the system was determined by numerically fitting the parameters of the equa-

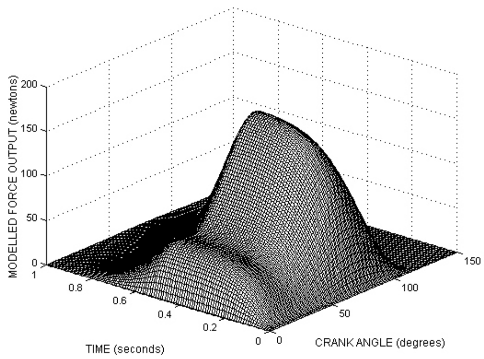

Fig. 7: The output force predicted by our model over crank angle and time

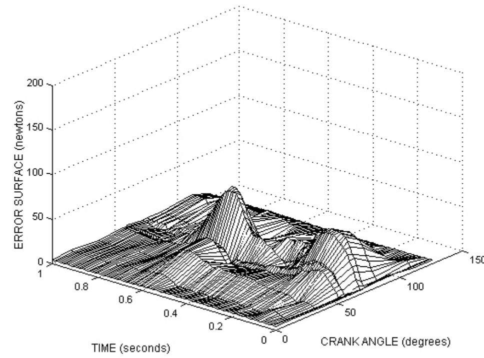

Fig. 8: The absolute error surface between model response and measured response. The peak errors (70newtons) occur during the force rise and fall sections of the response. The RMS error is 18.8newtons

tion to the measured response for an input of 10 stimulation pulses. It is given by

$$F_{step} = 1 - 3e^{-12t} + 2e^{-18t} \tag{3}$$

Using Laplace transforms the equation of motion of the force output due to stimulation of the L4L nerve root was derived as being

$$y'' + 30y' + 216y = \mathbb{N}\big(h_{L4L}(t)\big) \tag{4}$$

Next a solution for $f_{L4L}(t)$, the impulse response, was derived from the equation of motion as

$$f_{L4L}(t) = 360e^{-12t} - 360e^{-18t} \tag{5}$$

Using our model to predict the dynamic force response over a range of angles yields the surface in fig.7. This model response can be compared to fig.5 and is for the same stimulation level.

IV. Discussion

We have developed a model to approximate a complicated three-dimensional nonlinear system. The model fits the experimental data reasonably well in most areas. The surface describing the modelling error is given in fig.8. Improvement could be made in modelling the rate of change of the force output and we are currently working on adding a further block to the model to account for what appears to be a physical limiting factor.

V. Conclusion

Our model provides a first approximation to the complicated nonlinear dynamics found during stimulation of our subject's neuromuscular system and to our knowledge this is the first time that the dynamic force response of the leg has been measured and modelled for a spinal root stimulated human subject. Modifications should however be made to the model to account for the other non-linear effects which have been described here and work is continuing to try to develop our understanding of these effects.

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